

Winter 1994

Reconstruction and enhancement of medical computed tomography images

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ABSTRACT

Reconstruction and Enhancement of Medical Computed Tomography Images

by

Manoj K. Pattathil

The aim of Computed Tomography is to obtain information regarding the nature of the materials occupying precise positions inside the body. But due to errors in the projection data, which arise due to errors in measurement, beam hardening, missing data and noise, the reconstructed image has artifacts which reduces its diagnostic value. Thus the image obtained has to be processed to remove these artifacts and to improve the quality of the image. This is achieved by iterative reconstruction along with different post-reconstruction processing. Iterative reconstruction is done by repeating the process of projecting the image and creating new projection data which are then back projected to obtain an image of better quality. The algorithms that do these reprojections and back projections remove artifacts caused by errors mentioned above, thus enhancing the diagnostic value of the final image. Also the images obtained may be enhanced by a variety of post-reconstruction processes such as image enhancement techniques. Depending on the method these enhancement techniques may compensate for effects of specific degradation processes or can selectively emphasize or suppress selected features of the image.

This study is a) to test the accuracy of X-ray computed tomography image reconstruction applying a modified reprojection method and b) to enhance the image using a modification of the median filter. The experimental results demonstrated the superior performance of the methods.

RECONSTRUCTION AND ENHANCEMENT OF MEDICAL COMPUTED
TOMOGRAPHY IMAGES

by

Manoj K. Pattathil

A Thesis
Submitted to the Faculty of
New Jersey Institute of Technology
in Partial Fulfillment of the Requirements for the Degree of
Master of Science in Biomedical Engineering

Biomedical Engineering Committee

January 1994

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This thesis is dedicated to my uncle, aunt, and parents

ACKNOWLEDGMENT

I would like to express my sincere gratitude to my advisor Dr. Frank Shih for his guidance and motivation throughout this research. I also would like to thank him for leading me through my masters program.

Special thanks to Dr. Swamy Laxminarayan for his help and advice which made it possible for me to reach higher standards. My thanks go to Dr. David Kristol for providing help in the technical documentation of this thesis.

Finally I would like to thank Masoud Majidi for his timely help and valuable suggestions.

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CHAPTER 1

INTRODUCTION

1.1 Literature review

There are many situations in the field of computed tomography image reconstruction and processing in which it is necessary to perform accurate iterative reconstruction or iterative attenuation correction. This relies on the method chosen for the back projection from projection profiles to the reconstruction matrix, and the reprojection from the matrix to the pseudoprojection profiles. A number of efficient back projection and reprojection algorithms have been proposed which reduce total processing time by simplifying the process of interpolation used in determining the actual value to be back projected [1]-[5], [7]. A basic problem common to all of these endeavors is the accuracy with which the line integrals can be calculated from a finite field of pixels. Though in the case of the bone correction it would appear that an accurate algorithm is not essential because the correction made is relatively small in magnitude, it has been reported [6] that, in attempting to reduce image noise by using the least squares technique, the dominant source of noise becomes errors in the reprojecting process. Although many reprojecting algorithms have been proposed there is still a need for faster and more accurate algorithms especially to run on more efficient hardware becoming available today.

Since computed tomography and other noninvasive diagnostic imaging techniques are widely used because of their clinical image quality it has become increasingly important that along with faster processing time the accuracy of the image produced must be as high as possible, enabling the physician to make a reliable diagnosis, which is of utmost importance in surgery and its planning.

Thus many image enhancement methods have been proposed [8]-[12]. Though these methods are standard image enhancement techniques [12] they have been modified to suit the type of medical image that needs processing and also they depend on the type of errors to be corrected. A modified median filter method was developed in [8] which the edge is preserved and noise is reduced.

1.2 Introduction

1.2.1 Image Reconstruction

The problem of digitally reconstructing an image from its projections has become increasingly important during the past few years. There are many areas where practical applications of this problem arise. The application that has revolutionized diagnostic radiology is that of computerized tomography where X-rays are used to generate the projection data for a cross section of the human body. From the projection data an image is reconstructed of a cross section, depicting with very high resolution the morphological details of the body. Other medical applications are in the areas of nuclear medicine and ultrasonic imaging.

1.2.2 Image Enhancement

Image enhancement is done whenever the quality of the image obtained is not acceptable i.e. when the user is unable to perceive the important information is masked due to degradation processes caused by specific processes or due to inherent limitations of the imaging system. Thus many enhancement techniques are designed to compensate for specific degradation processes. They include methods of modifying the gray scale, deblurring, smoothing or removing noise, and correcting geometrical distortions. Increasing the contrast is a reasonable enhancement operation, since degradation usually attenuates the image signal;

deblurring is reasonable, since degradation usually blurs, and the original image is assumed to have been sharp. And smoothing is reasonable, since degradation usually introduces noise, and the original image is assumed to have been smooth. In general, image enhancement techniques are either used to suppress selected features of an image, or to emphasize such features at the expense of other features. Thus enhancement can be regarded as selective emphasis and suppression of information in an image.

CHAPTER 2

IMAGE RECONSTRUCTION

2.1 Introduction

The problem of image reconstruction from projections has arisen independently in a large number of scientific fields. An important version of the problem in medicine is that of obtaining the density distribution within the human body from multiple X-ray projections. This process is called computerized tomography, which has revolutionized diagnostic radiology. Images of cross sections of the human body are produced from data obtained by measuring the attenuation of x-rays along a large number of lines through the cross section. These data are called the projection data. The process of reconstruction produces an image of the cross section from the projection data. The reconstructed image obtained after the reconstruction process is generally not upto the clinical quality required for diagnostic purposes so post-reconstruction processes are applied to this image. This involves reprojecting the image to create a new set of projection data called the reprojection data which are then used to reconstruct the image after undergoing other image processing processes to correct artifacts caused by bone induced beam hardening, missing data, and noise. There are many algorithms for calculating projections which can be classified into two classes: 1) pixel intercepting methods, and 2) forward projection methods. Here a modified algorithm of the forward projection method is developed and experimentally evaluated using actual CT images.

2.2 Area Weighted Convolution Technique

This is a data reprojection algorithm developed for use in single photon emission computed tomography. The algorithm makes use of an accurate representation

of the pixel activity. The pixel is a square pixel of uniform intensity. The ability to perform accurate iterative reconstruction or iterative attenuation correction in single photon emission computed tomography (SPECT) relies fundamentally on the methods chosen for back projection from projection profiles to the reconstruction matrix, and reprojection from the matrix to the pseudoprojection profiles. But ultimately, the clinical image quality is limited to that which can be achieved in a processing time frame that is clinically reasonable. Thus the requirement of techniques that are both accurate and time efficient are basic in computed tomography.

This algorithm describes a method that uses the angle-dependent projection of the area of a square pixel in the reconstruction matrix to perform interpolation for the image reprojection. The algorithm has been called area weighted convolution (AWC) interpolation since it uses convolution of the projected pixel area with a subdivided projection array as the interpolation step.

2.3 Method

In the AWC algorithm each projection bin of width k is subdivided into n sub-bins of width k/n . This is to perform the pixel driven nearest neighbor projection on the subdivided projection. That implies that the pixel intensity value of each pixel is moved to the projection sub-bin whose center is closest to the intersection of the perpendicular from the center of the pixel to the projection array. The pixel driven technique is such that each pixel is operated on only once for the entire projection.

The AWC algorithm is based on the assumption that the pixel intensity is distributed uniformly throughout the pixel. Thus the distribution of a square pixel projected at an angle θ is the continuous set of line integrals of that pixel along the lines normal to the projection array. As shown , this distribution

function varies from a rectangular function of width k at even multiples of 45 degrees, to a triangular function of width $1.4k$ at odd multiples of 45 degrees. A trapezoidal function is obtained in between these two extremes. The area of each function is equal to the area of the square pixel (unity). Since these functions are symmetric about 45 degrees, they need only be calculated for one eighth of the total number of projection angles when uniformly sampled over 360 degrees.

The projection functions are then convolved with the subdivided projections after reprojection. The convolution operation actually distributes the intensity from each pixel about the site to which it was projected, according to its actual distribution at that angle.

The calculation of the projected distribution of the square pixel is as described. Referring to figure 1 at any angle θ ($0 < \theta \leq 45$ degrees), the area enclosed by the two parallel lines labelled C1 and C2 intersecting the lower left hand and upper right hand corners, respectively, projects as a rectangular function of fractional intensity F_1 . Any other projected fractional area F' is determined by calculating the width of the triangular portion and then linearly interpolating between the subbin with the maximum height to the subbin whose projection area is zero. For this the maximum pixel projection width is calculated as

$$\text{max width} = \sqrt{2} \times \cos(45 - \theta)$$

The method is implemented for the linear pixel dimension of unity. Thus the maximum width of the pixel projection possible is 1.41 times the pixel dimension which is 1.41. Hence to obtain an accurate pixel projection array the array was made 15 elements long. At angles less than 45 degrees it is evident that the pixel projection width is less than 1.41 and hence to obtain an accurate pixel projection array the values must be centered in the array flanked by zeros beyond the projection on either side. This is achieved by calculating the overhang on either side as follows

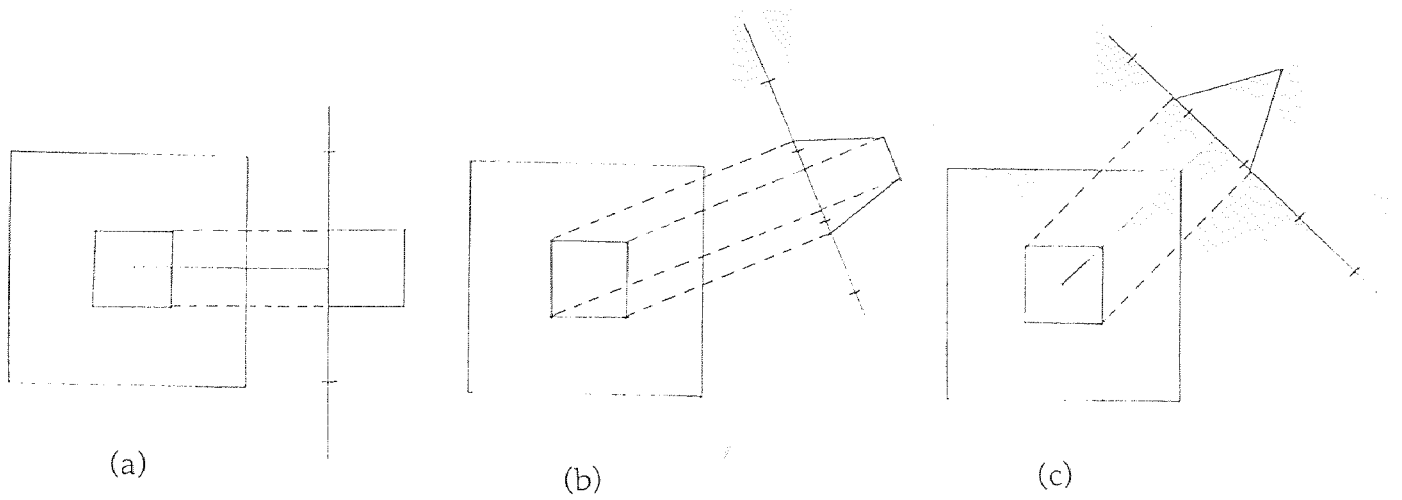


Figure 1 Projection of a uniform square pixel at (a) $\theta=0$, (b) $0<\theta<45$ deg, and (c) $\theta=45$ deg

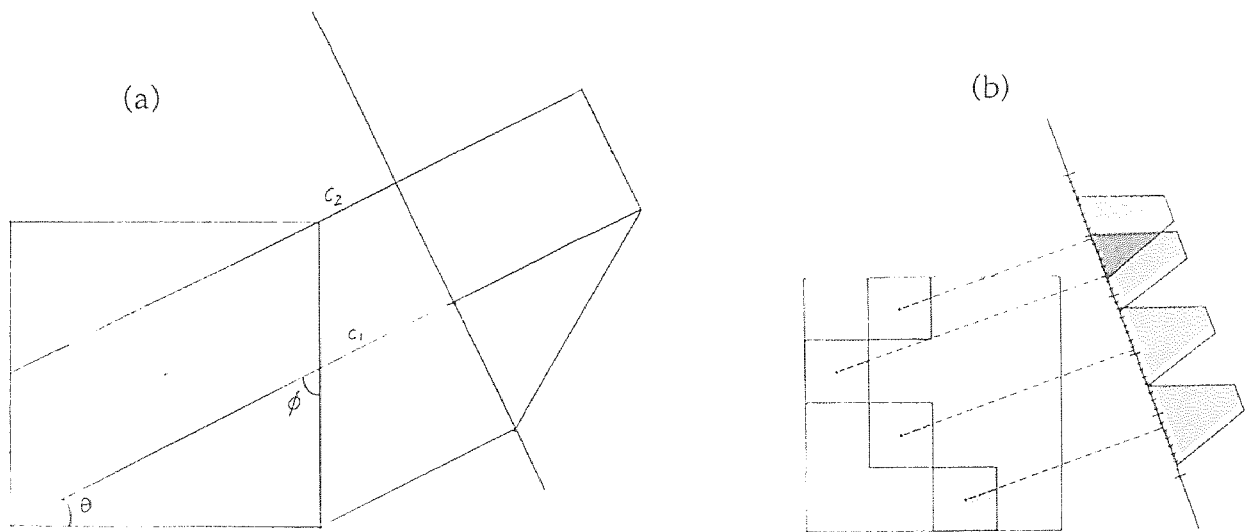


Figure 2 Projection of pixels (a) projection of a single pixel at angle θ (b) projection of several pixels onto the subdivided projection with their corresponding distributions.

$$\text{overhang} = (15 - \text{max width})/2$$

The area in the rectangular portion has a height given by

$$\text{max height} = \sec(\theta)$$

hence the area of the strips in the rectangular portion is given by $(0.1) \times (\text{max height})$. The area of the strips in the triangular portion is calculated as below:

$$\text{triangular portion strip area} = (0.1) \{ \sec(\theta) - 0.1 \} / \sin(\theta)$$

This has been implemented in the code inside a loop without the trigonometrical calculations, thus reducing computation time.

To perform the convolution of the projected pixel area distribution calculated as above with the subdivided projection array the subdivided projection array is convolved with the projected pixel's distribution array. The figure 2 shows the nearest neighbor reprojection of several pixels onto a subdivided projection array with their associated distribution functions as they would appear after the convolution step. This figure also represents the stage of the process just before compression of the projection array to the original bin size. The compression is done by simply summing up subbins between projection bin boundaries. This step enables the comparison of the reprojected image with its original measured projections, such as in iterative attenuation correction schemes or for iterative reconstruction. The bin size in each set of projections to be subtracted must be the same in all these applications.

CHAPTER 3

IMAGE ENHANCEMENT

3.1 Introduction

There are among many two particular approaches in enhancing morphological features in CT images. One is to increase the contrast of the suspicious areas and the other is to remove background noise. Methods of contrast enhancement are based on adaptive neighborhood processing with a set of contrast enhancement functions to enhance the features of the image. But in the method applied here the approach to enhance the images by removing background noise is taken. This is done while simultaneously preserving the edge information of the suspicious areas in the image.

3.2 Selective Median Filtering

Median filtering has been found to be very powerful in removing noise from two-dimensional signals without blurring the edges. This makes it particularly suitable for enhancing images. Median filters have several properties that make them superior to low-pass filters. If the image has impulse-like noise, median filtering can remove it without significantly distorting the signal, and if the image contains edges, median filtering can preserve them due to the fact that only a small fraction of the neighborhood overlaps the edge. Further, it has been found that the median filter is nearly optimal for suppressing impulse noise or noise which is characterized by a large percentage of outliers. Furthermore since the median filters can be defined for arbitrary sizes and shapes making them suitable for any kind of image.

3.3 Method

Since it was found that the edge preservation of the standard median filter was not sufficient for enhancing medical images due to the fuzziness of the boundaries of suspicious areas a modification of the median filter, the selective median filter (SMF), is used here for enhancement. To apply median filtering to a digital image, the value at each pixel is replaced by the median of the values in the neighborhood of the pixel. Given a set of n numbers $\{x_1 \dots x_n\}$ and the ordered set being defined as $\{x_1^* \dots x_n^*\}$. The median of the set is given by median of $\{x_1 \dots x_n\}$ is $x_n^*/2$ when n is even or is $(x_{n+1}^*)/2$ when n is odd

SMF is defined as follows:

For a window $W(i,j)$ centered at image coordinates (i,j) , the output of the selective median filter is:

$SMF(x_{ij}) = \text{median} \{ x_{r,s} \text{ such that } (r,s) \text{ belongs to } N(i,j) \text{ and } |x_{rs} - x_{ij}| < T \}$, where (i,j) belongs to the image Z^2 .

$N(i,j)$ is the area in the image covered by the window $W(i,j)$ and T is a threshold.

In computing the median, the set of pixels is restricted to those with a difference in gray level no greater than some threshold 'T'. If T is small the edge preservation of SMF is strong, but its smoothing function will be small. SMF behaves the other way if T is large.

To achieve strong noise suppression a filter with a large window size may be used or else the filter can be applied repeatedly. But it has been shown that edge information is not lost by iterative filtering and hence is more suited for application on medical images.

CHAPTER 4

RESULTS

4.1 Image Reconstruction

The new area weighted convolution reprojection algorithm was implemented in the C language on a Sun Sparc workstation for a parallel beam geometry. The new algorithm was implemented for a bin subdivision factor of 10.

The accuracy of the new algorithm was tested using computed tomography images of the transverse sections of the head and the vertebra, and longitudinal section of the heart. The images used were of sizes 128x128 and 256x256 pixels. The different sizes and intensities of the images along with their shapes allowed for an effective test of the accuracy of the algorithm. The algorithm was implemented for a parallel beam geometry. A total of 72 projections were evaluated over 360 degrees with uniform angular sampling. The projection data were calculated for 0 to 180 degrees and was duplicated for 180 to 360 degrees. The projection bin width was set equal to the linear pixel dimension.

The projection data was then used to reconstruct the image using the back projection function of the Viewit medical imaging software. Viewit is a software developed by the National Center for Supercomputing Applications at the University of Illinois, and can be described as a scientific calculator for multidimensional arrays with extra buttons for signal processing, scientific visualization, image processing and multivariate analysis. This back projection function performs a parallel back projection of a set of projections defined as a two dimensional array in the ss1 stack register using the acquisition geometry defined as a one dimensional vector of theta values or the projection angles in the ss2 register. The theta values are in radians. Thus the implementation of the

algorithm returned the projection data and the corresponding theta value. The linear interpolation method used for the backprojection. The images were reconstructed to the same size as that of the original so as to verify the accuracy of reconstruction.

The figures show the original computed tomography images and the figures show the reconstructed images. It is seen that all image details have been accurately reproduced. The exterior of the reconstructed images show negligible errors and even the artifacts have been faithfully reconstructed. In the case of the vertebra (figures 3 and 4) the narrow bone structure is reproduced almost accurately showing that the algorithm is effective even for narrow structures. The disc has in fact been enhanced and details are clearer. In the case of the heart image (figures 5 and 6) it is seen that there is a slight modification of the intensity but all the details are preserved. It is also noted that the artifacts have been reproduced identically. The head slice (figures 7 and 8) is a composite of bone and tissue and serves as an excellent test to more accurately judge the performance of the algorithm. It is seen that the bony structure has been accurately reproduced and the tissue structure which forms the less intense portion of the image shows all details. Though it may be noted that there is a negligible amount of smoothing at sharp edges. This is seen only when the intensity contrast is great. This problem may be overcome by using more subbins per projection bin which enables a finer sampling of the square pixel distribution and also by increasing the number of projection angles.

4.2 Image Enhancement

The selective median filter algorithm for enhancing images was implemented in C language on the Sun Sparc workstation. The chest radiographs of size 512x512 were used to evaluate the performance of the algorithm. The selective median

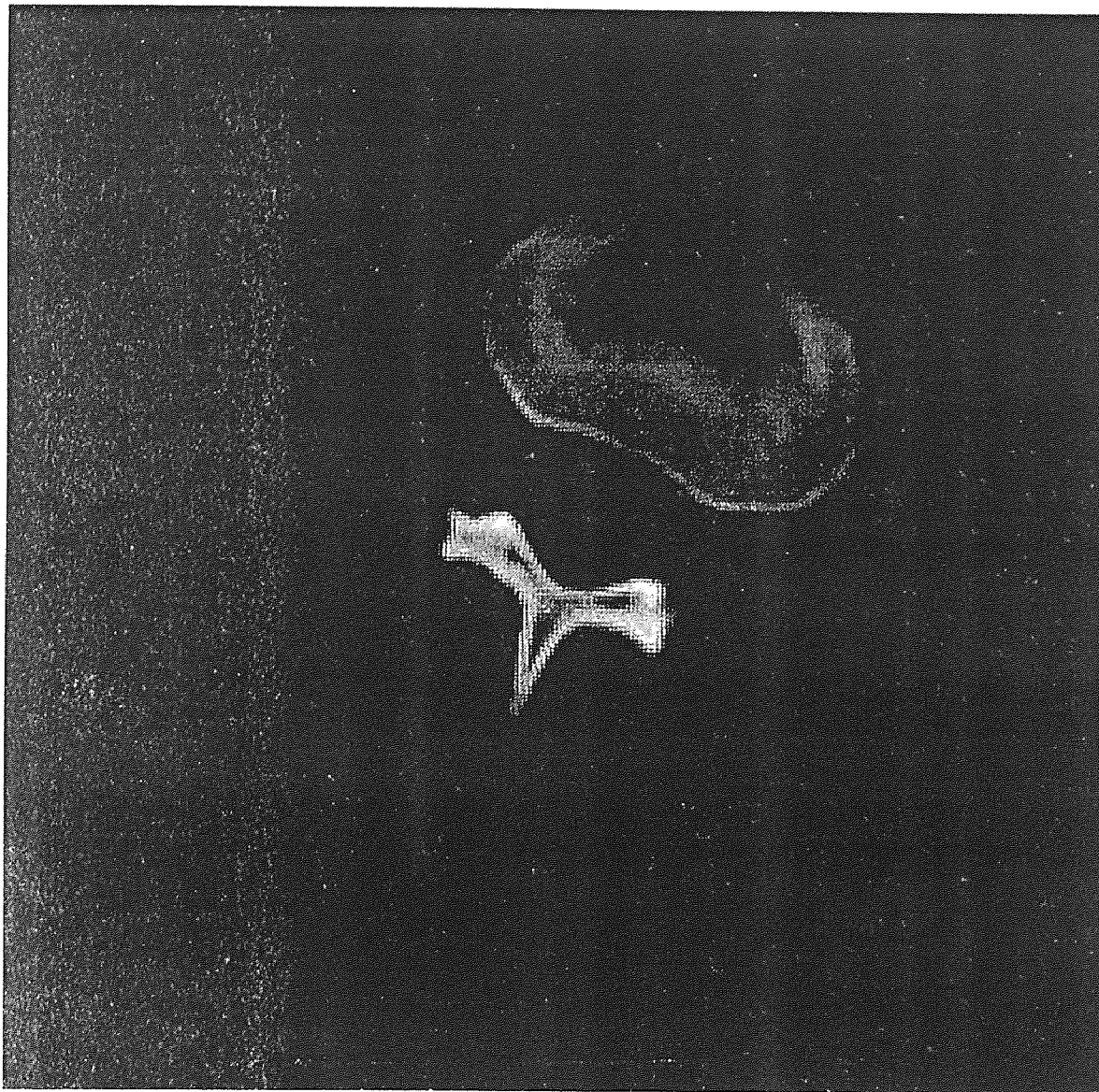


Figure 3 C T image of transverse section of the vertebra

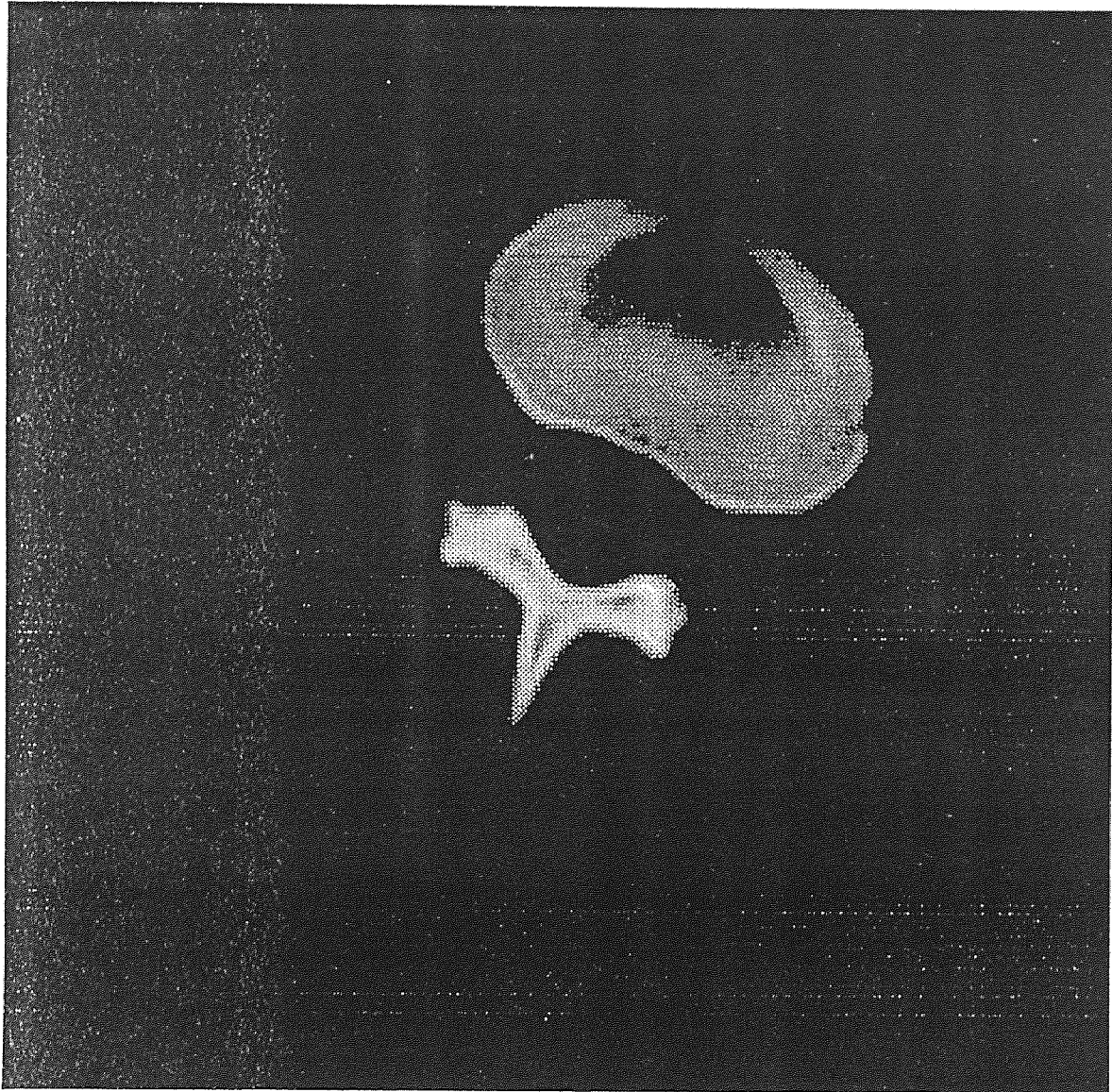


Figure 4 Reconstructed image of the vertebra

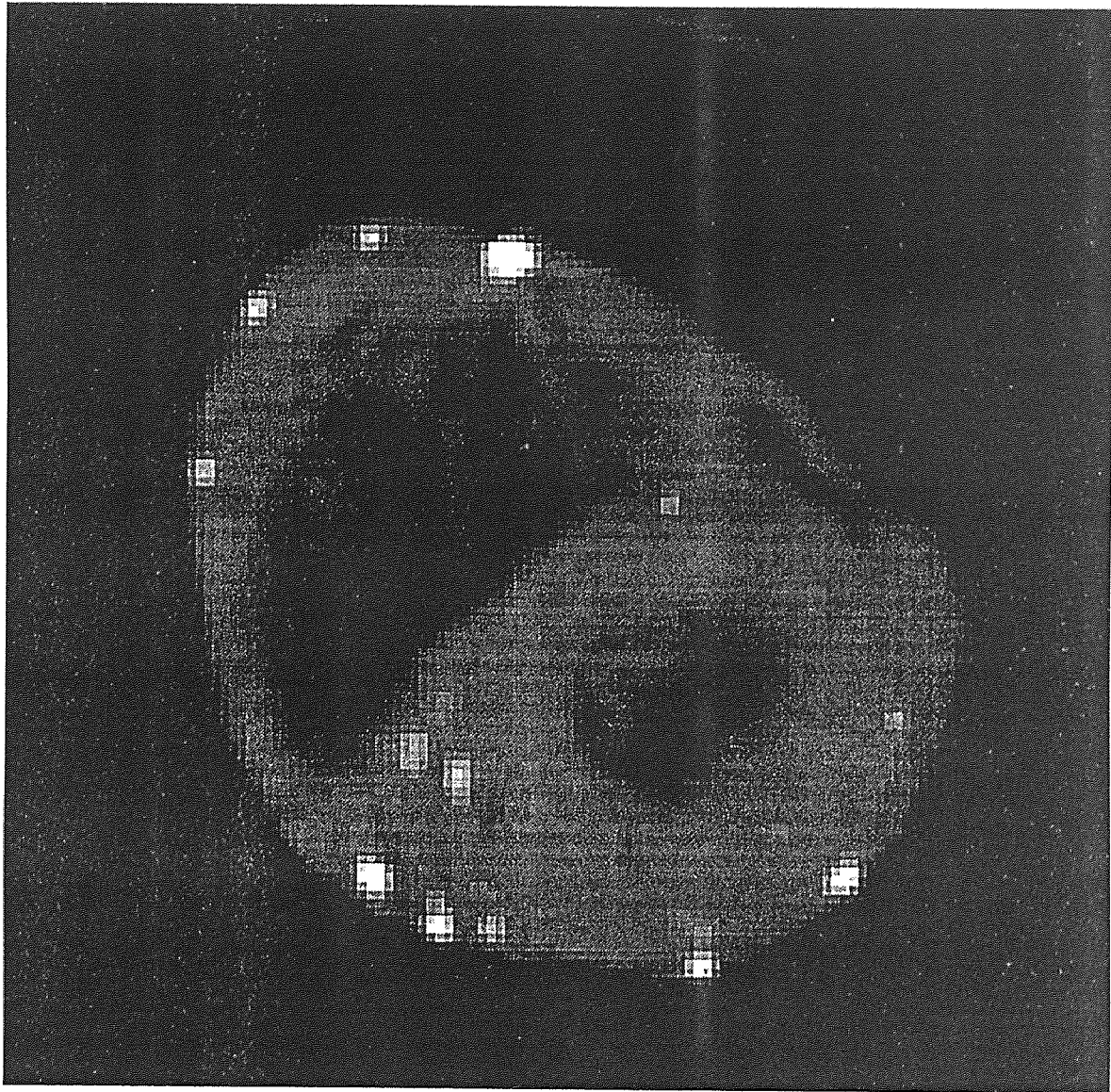


Figure 5 C T image of the transverse section of heart

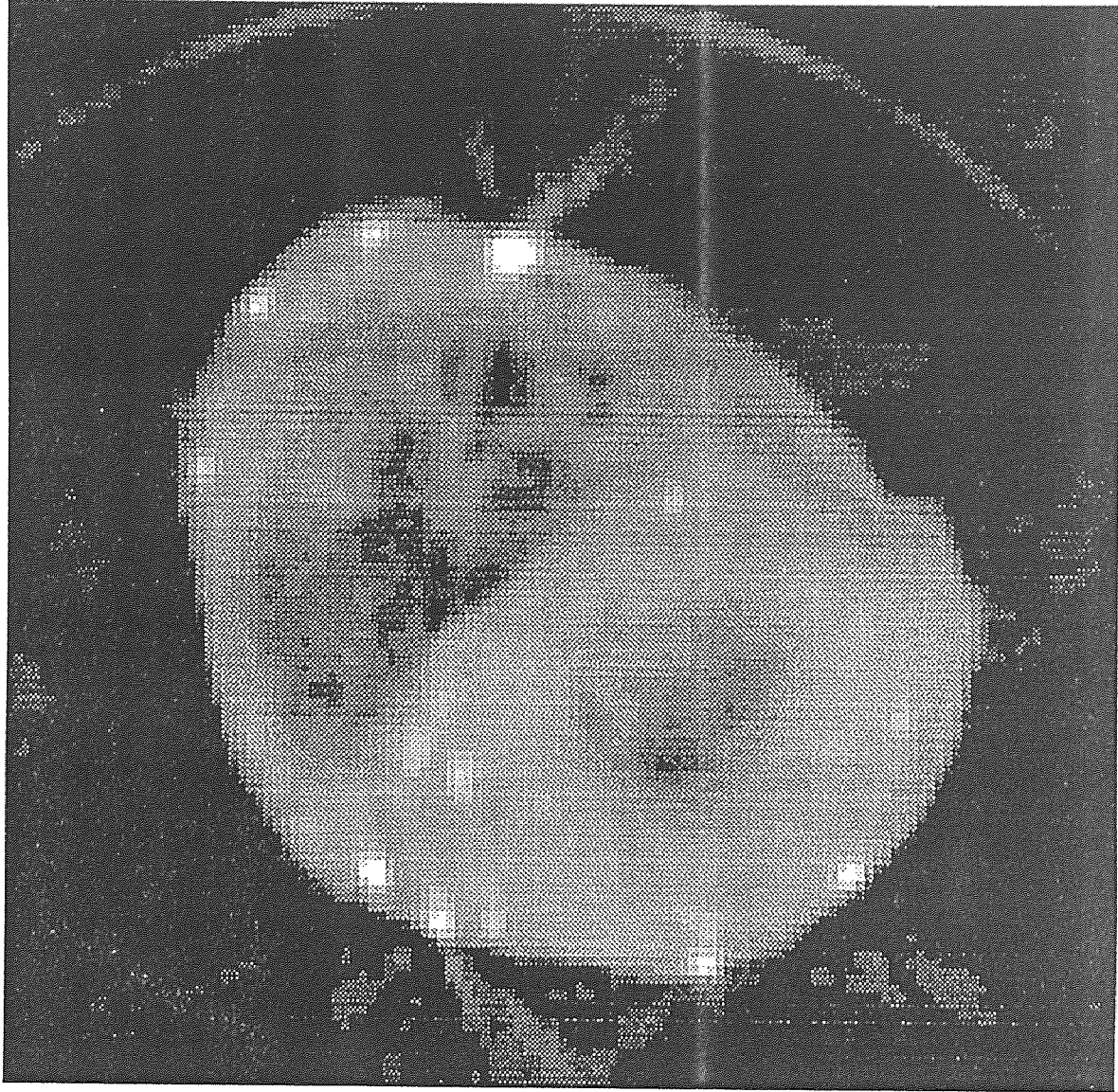


Figure 6 Reconstructed image of the heart

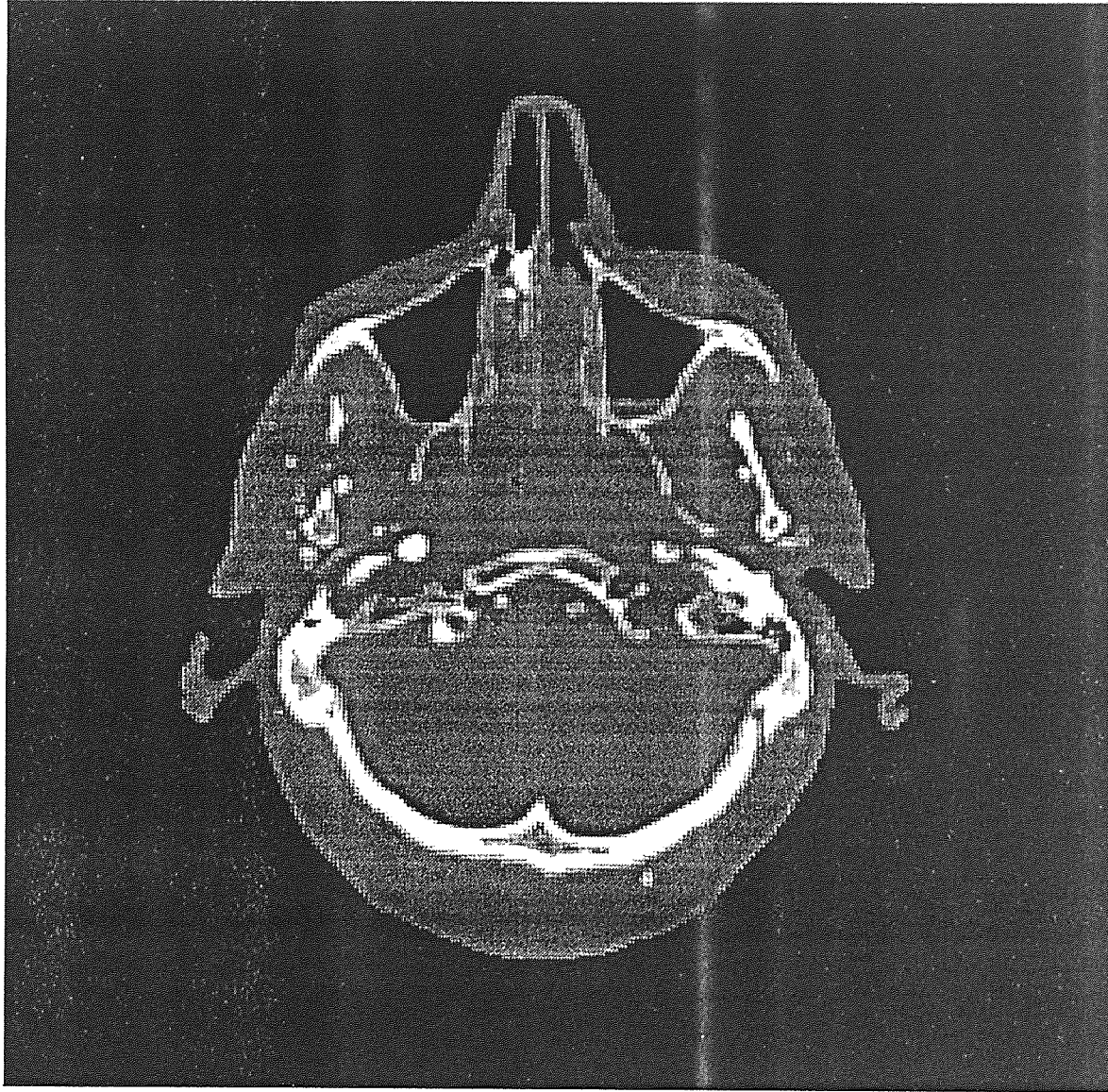


Figure 7 C T image of the transverse section of the head

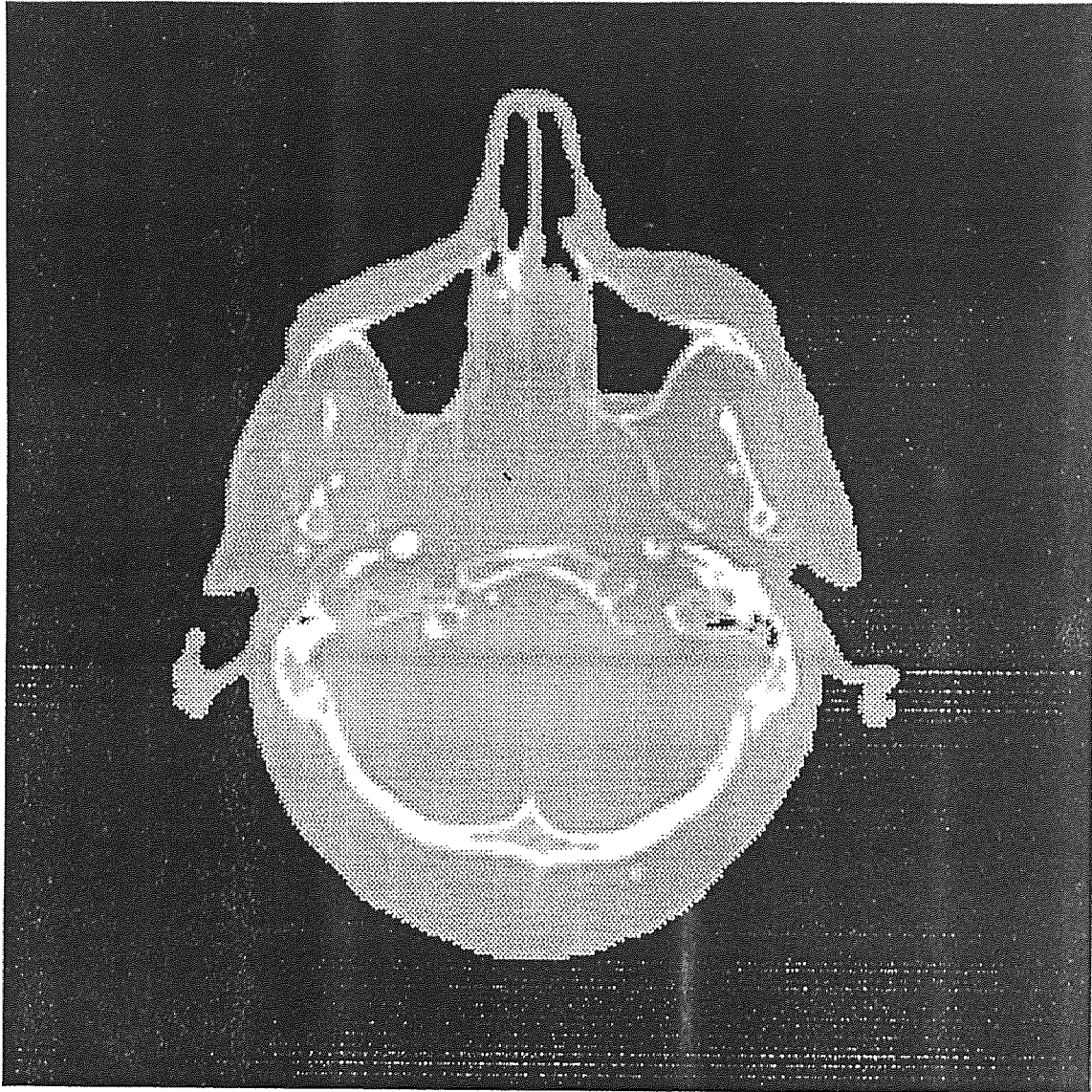


Figure 8 Reconstructed image of the head

filter has two parameters that can be adjusted to the noise characteristics of the image, the parameters being the threshold and the window size. To estimate the threshold the contrast between pixels in the image was checked. It was noted that the contrast ranged from 5 to 15 gray levels. So using different threshold values within this range, the filter was applied to the images window sizes of 5,9,and 13.

It was noted that as the window size was increased the noise suppression increased but it also caused signal distortion. And with a smaller threshold the edge preservation was strong, but increasing the threshold increased the smoothing which reduced the edge preservation. Thus it was concluded that for the images tested the best performance was obtained for a window size of 9x9.

The figures 9 and 13 are the test images of gray level range of 0 to 255. The selective median filter algorithm was applied to these images with a window size of 9x9 and threshold values of 5, 7 and 10. It is seen from figures 10, 11, 12, 14, 15, and 16 that the edge preservation of the filter decreases with the increase of the threshold and thus the details of the output images are reduced. Compared to the original images the background variation is reduced, while the boundaries are preserved.

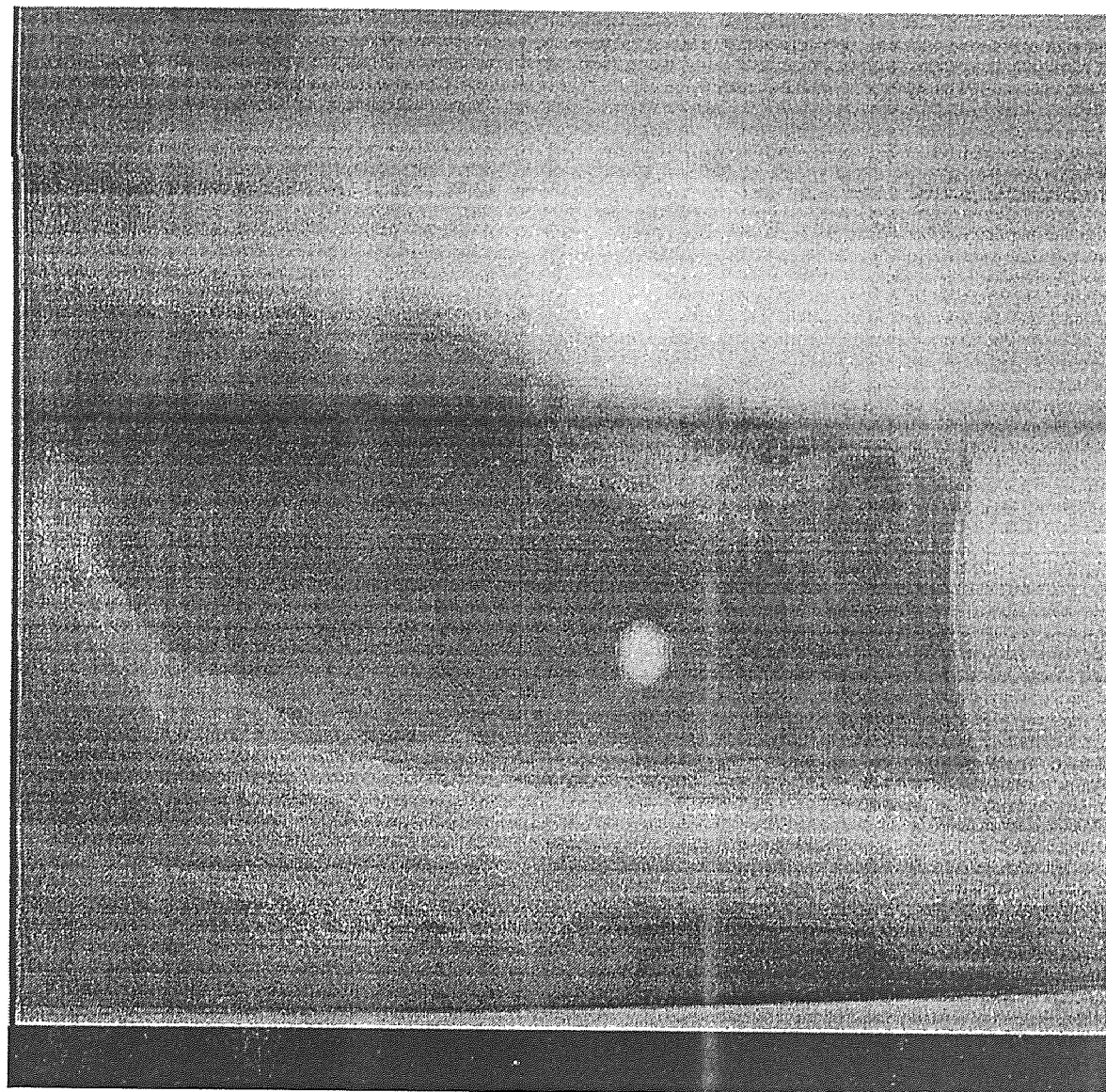


Figure 9 Original chest x-ray image (1)

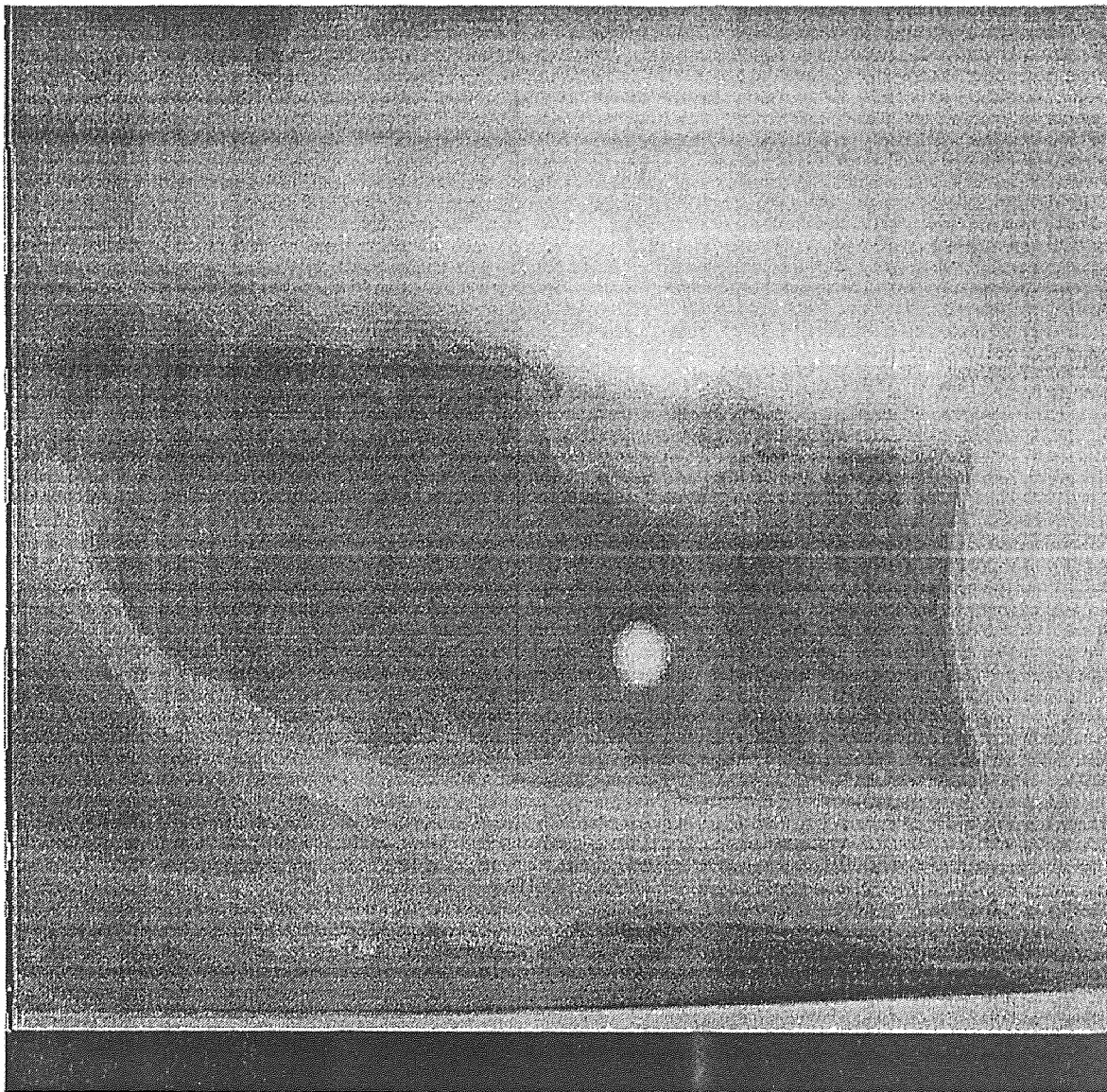


Figure 10 Enhanced image after applying SMF with $T = 5$

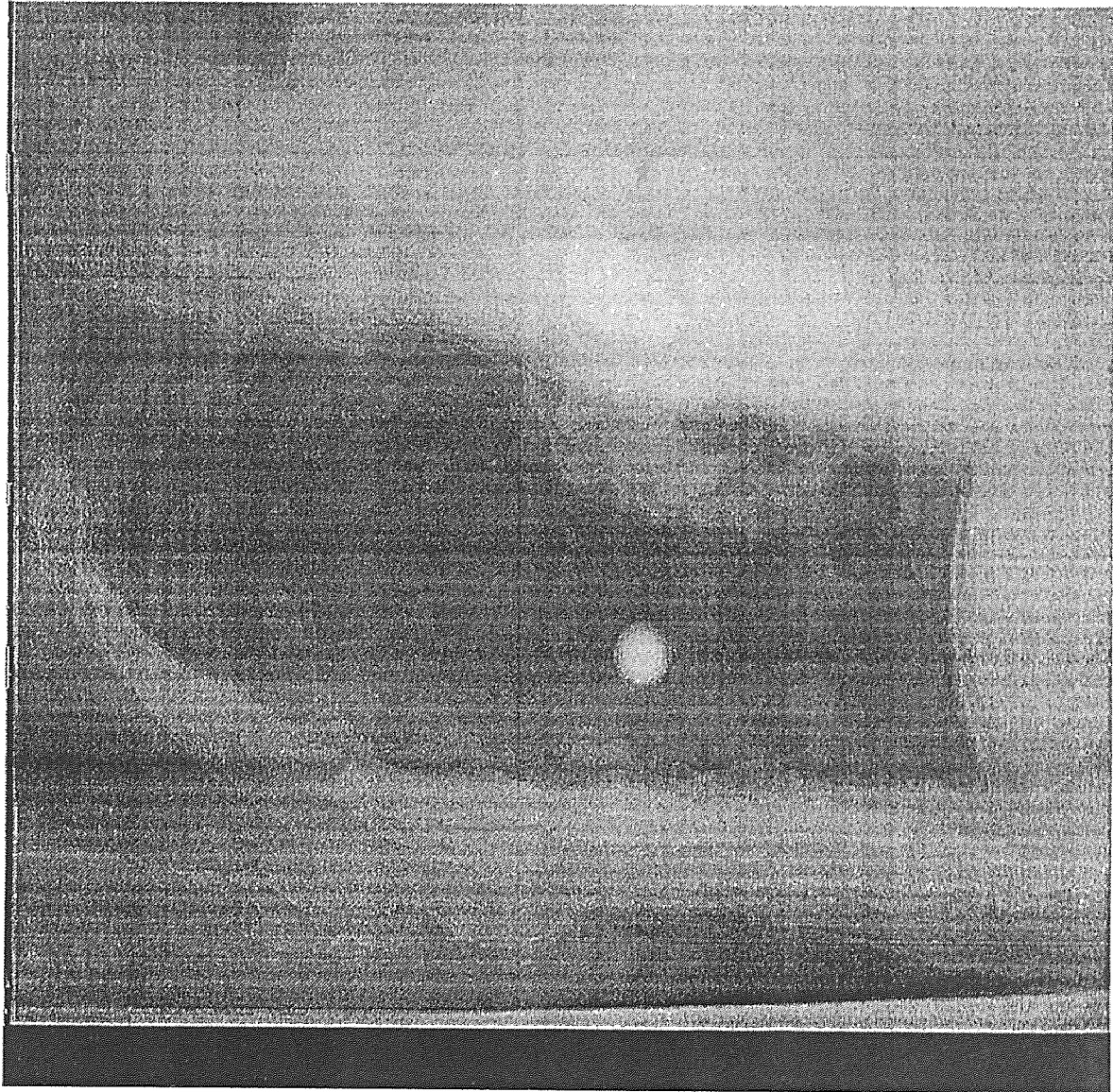


Figure 11 Enhanced image after applying SMF with $T = 7$

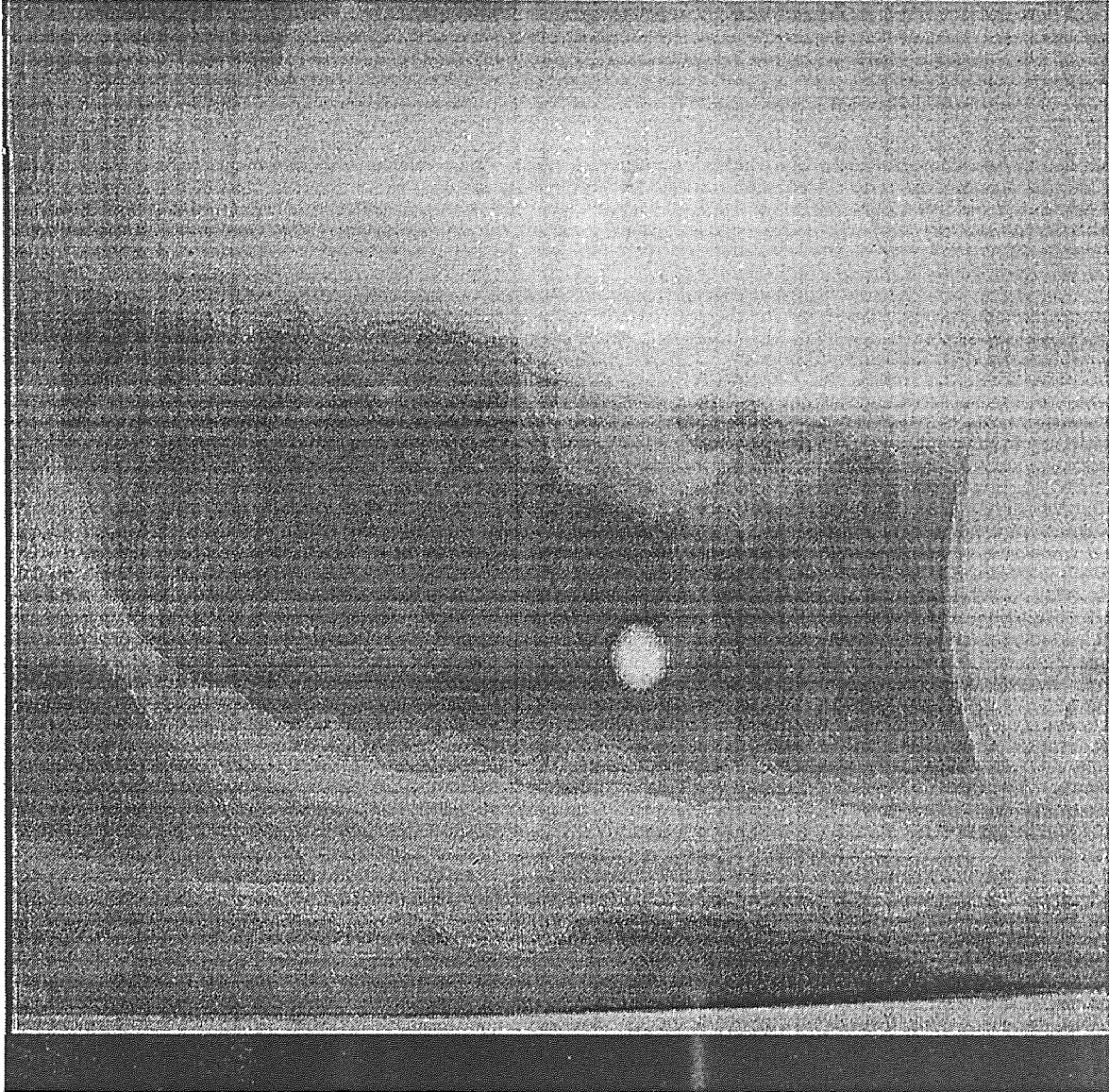


Figure 12 Enhanced image after applying SMF with $T = 10$

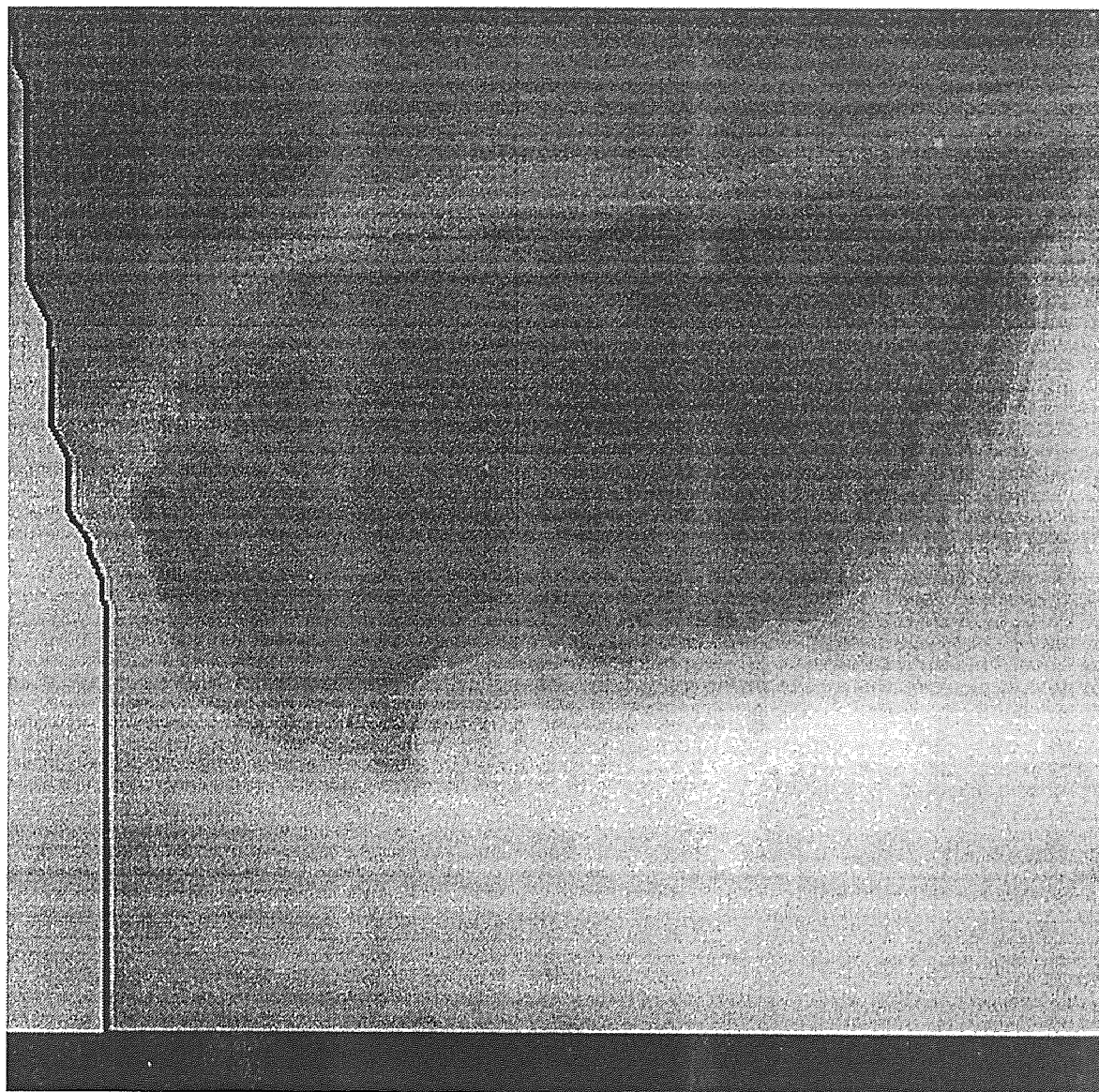


Figure 13 Original chest x-ray image (2)

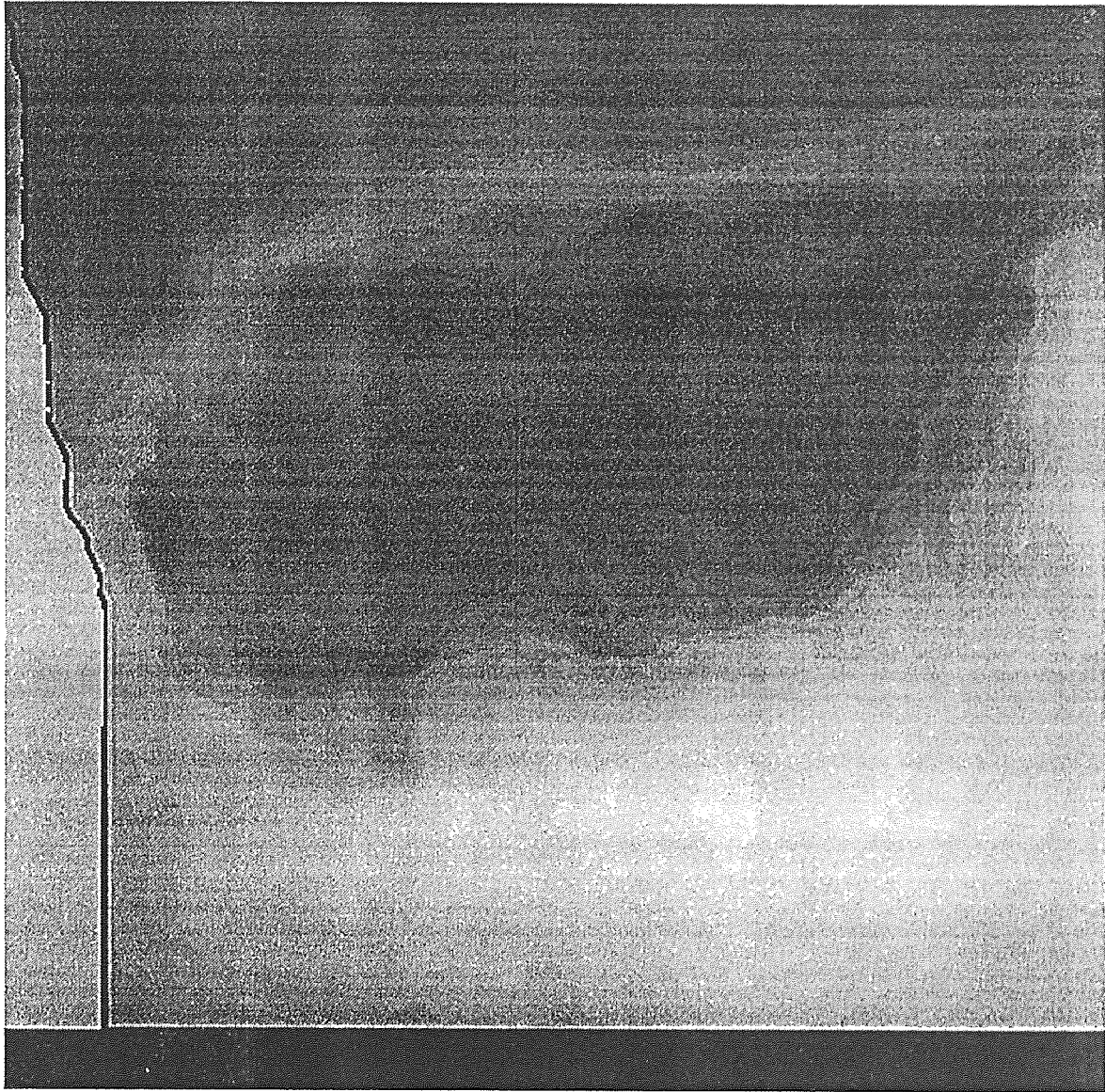


Figure 14 Enhanced image after applying SMF with $T = 5$



Figure 15 Enhanced image after applying SMF with $T = 7$



Figure 16 Enhanced image after applying SMF with $T = 10$

CHAPTER 5

CONCLUSIONS

5.1 Discussion

The new area weighted convolution algorithm of interpolation for reprojecting data from reconstructed computed tomography images has been shown to have a very good accuracy. Furthermore this method is capable of being easily implemented on array processors, thus capable of very fast execution times, which is essential with the faster array processors becoming available. Also the method of implementation described here makes this algorithm quite suitable for back projection. Thus its use as a fast and accurate back projection and reprojection pair would be of benefit in iterative reconstruction and attenuation correction of computed tomography images.

The enhancement of images using the selective median filter algorithm was shown to be effective in preserving the edges while removing noise. This property of the selective median filter is especially suited for defect detection applications in medical imaging especially when the original image quality is too poor to be of adequate diagnostic value. Thus an excellent field of application of this method is in evaluation of mammograms for cancer detection.

5.2 Future research

There are numerous directions in which these image processing techniques can be applied and suitably modified. One of the most promising field of application is in the computerized mammogram screening to detect tumors. Breast cancer is a leading cause of death among all cancers for women of middle age and older, and primary prevention is not possible as the cause of the disease is not

completely understood. However, current methods of treatment are very effective in the early phase of the cancer. Thus for effective treatment the tumorous growth should be detected in the early stages for which mammography is the most reliable method. But the mammograms have to be screened, and due to the shortage of trained radiologists the computerized screening will offer a dramatic cost benefit along with the ability to handle large volumes of data. Hence the methods described can be adapted to computerized screening of medical images at large.

APPENDIX A

Area Weighted Convolution Reprojection

```
#include<stdio.h>
#include<math.h>
#define size 256
#define max_bin 15
#define epsilon 0.00001
#define inc 10.0
#define PI 3.1415926
main()
{
    int i,image[size][size],proj_array[7200];
    double angle,filter_array[15],awc_result[450];
    angle=0.0;
    for(i=0;(i<360)&&(angle<180.0);i++)
    {
        read_image("head_slice",image);
        filter(filter_array,angle);
        proj(image,angle,proj_array);
        convolve(filter_array,proj_array,awc_result);
        angle=angle+inc;
    }
}

/* Creates the projection array */

proj(img,angle,proj_array)
int img[size][size],proj_array[7200];
double angle;

{
    int i,j,subbin,proj_size,pix_val;
    double g,h,a,b,d,e,f,s,p,t,rangle,q,x1,y1,x2,y2,xprime,yprime,z;
    FILE *fp2;
    fp2=fopen("thetaray","w");

    rangle=angle*PI/180.0;
    fprintf(fp2,"%lf ",rangle);

    if(angle<=90.0)
    {
        t=angle*PI/180.0;
        p=(90.0-angle)*PI/180.0;
    }
    if((angle>90.0)&&(angle<180.0))
    {
        t=angle*PI/180.0;
        p=(angle-90.0)*PI/180.0;
    }
}
```



```

a=cos(p);
b=cos(t);
e=size/2.0;
f=size/sqrt(2.0);

for(i=0;i<size;i++){
    for(j=0;j<size;j++){
        {
            pix_val=img[i][j];
            yprime=e-((double)j+0.5);
            xprime=e-((double)i+0.5);
            s=(a*yprime+b*xprime);
            x1=(double)i+0.5+b*s;
            y1=(double)j+0.5+a*s;

            x2=e;
            y2=e;
            g=(x1-x2)*(x1-x2)+(y1-y2)*(y1-y2);
            h=sqrt(g);
            z=ceil(h*(double)max_bin);
            subbin=(int)z;

            proj_array[subbin]=proj_array[subbin]+pix_val;
        }
    }
    fclose(fp2);
}
/* Function reads in binary image file */

read_image(filename,img)
char filename[];
int img[size][size];
{
    int i,j;
    unsigned char c;

    FILE *fp;

    fp=fopen(filename,"r");
    if (fp==NULL)
    {
        printf("error\n");
        exit (-1);
    }

    for(i=0;i<size;i++)
        for(j=0;j<size;j++)
        {
            fread(&c,sizeof(char),1,fp);
            img[i][j]=c;
        }

    fclose(fp);
}
/*

```

```

for(i=0;i<size;i++)
for(j=0;j<size;j++)
    if((img[i][j]<0) || (img[i][j]>255))
        { printf("%d ",img[i][j]);
          exit(-1);
        }
*/
}

/* Function convolves the pixel projection with the projection array */
convolve(filter_array,proj_array,awc_result

int proj_array[7200];
double filter_array[max_bin],awc_result[450];

{
    int i,j,k,proj_size,over_hng;
    double proj_buf[7200];
    FILE *fp1;

    fp1=fopen("projdat","w");
    if(fp1==NULL)
    {
        printf("error\n");
        exit(-1);
    }

    over_hng=(int)floor(max_bin/2.0);
    proj_size=(int)ceil(size*sqrt(2.0)*max_bin);

    for(i=0;i<proj_size;i++)
        proj_buf[i]=0;

    for(i=over_hng;i<proj_size-over_hng;i++)
        for(j=0;j<max_bin;j++)
            proj_buf[i+j-over_hng]=proj_buf[i+j-over_hng]+filter_array[j]*proj_array[i];

    k=0;
    j=0;

    for(i=0;i<proj_size;i++)
    {
        awc_result[k]=awc_result[k]+proj_buf[i];
        j=j+1;
        if(j%11==0)
            k=k+1;
        fprintf(fp1,"%lf ",awc_result);
    }
    fprintf(fp1,"\n");
    fclose(fp1);
}

```

```

/* Function creates the pixel projection */

filter(filter_array,angle)
double filter_array[15],angle;

{
    int i,k,o,s,w, temp,max_div,filter_size;
    double p,q,t,filtang,sin_theta,cos_theta,rect_length,
    triang_area,filter_width;
    if(angle<=45.0)
        filtang=angle;
    else
        if((angle>45.0)&&(angle<=90.0))
            filtang=angle-45.0;
        else
            if((angle>90.0)&&(angle<=135.0))
                filtang=angle-90.0;
            else
                if((angle>135.0)&&(angle<=180.0))
                    filtang=angle-135.0;
                t=filtang*PI/180.0;
                filter_width=sqrt(2.0)*cos((45.0-filtang)*PI/180.0);
                s = (int) ceil(filter_width*10);
/*
    max_div = ((s % 2) == 0) ? s+1 : s;
    printf ("s = %d\t\tmax_div = %d\n", s, max_div);
*/
    sin_theta=sin(t);
    cos_theta=cos(t);
    rect_length=1.0/cos_theta;
    w=(int)floor((filter_width-2.0*sin_theta)*10.0);
    max_div=max_bin;
    o=(int)((max_div-s)/2);

    for(i=0;i<max_div;i++)
        filter_array[i]=0.0;

/* calculation of area distribution for the pixel projection */

    for(i=0;i<max_div;i++)
    {
        if((o>epsilon)&&(i<o))
            filter_array[i]=filter_array[max_div-i-1]=0.0;
        if((angle>epsilon)&&(i>=0))
        {
            /* calculation for rectangular region */
            if((filtang!=45.0)&&(i>=(max_div-w-o)/2)&&(i<max_div-(max_div-w-o)/2))
                filter_array[i]=rect_length;
            if((i<(max_div-w)/2)&&(i>o-1))
            {
                triang_area=rect_length * (i+1.0)/(1.0+(max_div-w-o)/2.0);
                filter_array[i] = filter_array[max_div-i-o] = triang_area;
            }
            if(filtang==45.0)

```

```
    {
        filter_array[(max_div-1)/2]=rect_length;
    }
    else
        if((i>o)&&(i<max_div-o))
        {
            filter_array[i]=rect_length;
        }
    }
    filter_size=max_div;
}
```

APPENDIX B

Selective Median Filter

```
#include<string.h>
#include<stdio.h>
#define NULL 0

int sort_array();

/* Routine to calculate median array in the window */

int sort_array(a,n,p)
int *a;
int n;
int p;
{
    int i,j,temp1,temp2;
    if (n==0) {
        a[0] = p;
        return (1);
    }
    else {
        for (i=0;i<n;i++) {
            if (p<a[i]) {
                temp1 = a[i];
                a[i] = p;
                for (j=i+1;j<n;j++){
                    temp2 = a[j];
                    a[j] = temp1;
                    temp1 = temp2;
                }
                a[n] = temp1;
                n++;
                return(n);
            }
        }
        if (p>a[n-1]) {
            a[n] = p;
            n++;
            return(n);
        }
    }
    return(n);
}

/* To calculate the selective median filter and apply
it to the image, returning the enhanced image */

main(argc,argv)
int argc;
```

```

char *argv[];
{
    unsigned char *image, *filter;
    FILE *fp;
    int i,j,r,s,n;
    static char ct[3] = ".c";
    static char cb[3] = ".y";
    char *ext;
    char file_name[25];
    int xsize,ysize,xstart,ystart,xend,yend;
    int arr[81];
    int T=5;

    /* Read in the image to be enhanced */

    strcpy(file_name,argv[1]);
    ext = strtok(argv[1],ct);
    xsize = atoi(strtok(NULL,ct));
    if ((ext=strtok(NULL,cb))!=NULL) ysize = xsize;
    else ysize = atoi(ext);

    image = (unsigned char *)calloc(xsize*ysize,sizeof(unsigned char));
    filter = (unsigned char *)calloc(xsize*ysize,sizeof(unsigned char));

    if ((fp = fopen(file_name,"r"))== NULL) {
        printf("Could not open the file\n");
        exit(1);
    }

    fread(image,sizeof(unsigned char),xsize*ysize,fp);
    fclose(fp);

    /* Apply the Selective Median Filter and store the results in "decode" */

    for (j=0;j<ysize;j++) {
        for (i=0;i<xsize;i++) {
            if ((j-4)<0) ystart = 0;
            else ystart = j-4;
            if ((i-4)<0) xstart = 0;
            else xstart = i-4;
            if ((j+4)>ysize) yend = 0;
            else yend = j+4;
            if ((i+4)>xsize) xend = 0;
            else xend = i+4;
            n = 0;
            for (s=ystart;s<yend;s++) {
                for (r=xstart;r<xend;r++) {
                    if (abs(image[s*xsize+r]-image[j*xsize+i])<T){
                        n=sort_array(arr,n,image[s*xsize+r]);
                    }
                }
            }
            if (n%2==0) filter[j*xsize+i] = arr[(n/2)];
        }
    }
}

```

```
        else filter[j*xsize+i] = arr[(n/2)+1];
    }
}
fp = fopen("decode","w");
fwrite(filter,sizeof(unsigned char),xsize*ysize,fp);
fclose(fp);
}
```

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